Gait & Posture 51 (2017) 41–46

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

Medial-lateral centre of mass displacement and base of support are equally good predictors of metabolic cost in amputee walking

R.A. Weinert-Aplin, PhD[a,*], M. Twiste, PhD[a,b], H.L. Jarvis, PhD[a,c], A.N. Bennett, PhD FRCP[c,d,e], R.J. Baker, PhD[a]

[a] School of Health Sciences, University of Salford, Salford, UK
[b] United National Institute for Prosthetics & Orthotics Development, UK
[c] Defence Medical Rehabilitation Centre Headley Court, Surrey, UK
[d] Leeds Institute of Rheumatic and Musculoskeletal Medicine, University of Leeds, UK
[e] National Heart and Lung Institute, Faculty of Medicine, Imperial College London, UK

ARTICLE INFO

Article history:
Received 26 January 2016
Received in revised form 19 September 2016
Accepted 26 September 2016

Keywords:
Efficiency
Gait
Prosthetic
Rehabilitation

ABSTRACT

Amputees are known to walk with greater metabolic cost than able-bodied individuals and establishing predictors of metabolic cost from kinematic measures, such as centre of mass (CoM) motion, during walking are important from a rehabilitative perspective, as they can provide quantifiable measures to target during gait rehabilitation in amputees. While it is known that vertical CoM motion poorly predicts metabolic cost, CoM motion in the medial-lateral (ML) and anterior-posterior directions have not been investigated in the context of gait efficiency in the amputee population. Therefore, the aims of this study were to investigate the relationship between CoM motion in all three directions of motion, base of support and walking speed, and the metabolic cost of walking in both able-bodied individuals and different levels of lower limb amputee. 37 individuals were recruited to form groups of controls, unilateral above- and below-knee, and bilateral above-knee amputees respectively. Full-body optical motion and oxygen consumption data were collected during walking at a self-selected speed. CoM position was taken as the mass-weighted average of all body segments and compared to each individual's net non-dimensional metabolic cost. Base of support and ML CoM displacement were the strongest correlates to metabolic cost and the positive correlations suggest increased ML CoM displacement or Base of support will reduce walking efficiency. Rehabilitation protocols which indirectly reduce these indicators, rather than vertical CoM displacement will likely show improvements in amputee walking efficiency.

1. Introduction

It is known that lower limb amputees walk less efficiently than able-bodied individuals, with progressively worse efficiency as the level of amputation increases [1–4]. To assess walking, and in particular walking efficiency in lower limb amputees, a range of biomechanical and physiological parameters have been used, including Centre of Mass (CoM) displacement and various respiratory measures [5]. Specifically, the respiratory measure considered most related to walking efficiency is the metabolic cost of walking and has been used to assess over-ground and treadmill walking [6–8] when comparing between able-bodied individuals or between amputee groups [1,2,9–11] or between different prosthetic devices within amputee groups [12–15]. To avoid confusion, this study considers more efficient gait to be when the metabolic cost, defined as the metabolic energy expended to move a unit distance, decreases.

As it is not always possible to obtain metabolic data, studies have sought to establish other predictors of the cost of walking, such as walking speed [16] or vertical CoM displacement [17,18]. This follows the work of Saunders et al. [19] who presented the six determinants of gait which were seen to influence CoM motion, the main biomechanical parameter historically believed to be related to the energetic cost of walking. This idea was based on the observation that pathological gait deviated from what was considered "normal". In particular, the observed greater CoM displacements in pathological gait suggested more mechanical work was being performed compared to a "normal" gait pattern,

* Corresponding author at: PO33, Brian Blatchford Building, University of Salford, Salford, M6 6PU, UK.
E-mail address: r.a.weinert-aplin@salford.ac.uk (R.A. Weinert-Aplin).

http://dx.doi.org/10.1016/j.gaitpost.2016.09.024
0966-6362/

and therefore more energy would be required to achieve this. While excessive CoM motion in the medial-lateral and anterior-posterior directions were also considered undesirable, the focus of these determinants tended to be on avoiding excessive vertical CoM displacement. This is somewhat simplistic in that displacements of the CoM in a vertical direction allow for an interchange between kinematic and potential energy which almost certainly reduce the requirement for work to be done and thus the metabolic cost. While it is reasonable to assume excessive vertical displacement would be indicative of increased cost of walking, there is no obvious reason to assume minimising it would minimise energy cost. Recent studies have indeed shown that deliberately reducing CoM motion actually increases metabolic cost [17,18]. Studies have also shown that several of the determinants make negligible difference to CoM motion [20–22]. Additionally, the determinants have recently been assessed in the context of inverted pendulum walking [23], and have found the major cost of walking was attributed to redirecting the CoM during the step-to-step transitions [24]. There have been few studies attempting to relate metabolic cost of walking to biomechanical factors in people with pathologies but the most comprehensive suggested that vertical centre of mass excursion was not a good indicator of metabolic cost in people with myelomeningocele [25].

In addition to vertical CoM displacement and sagittal plane measures of walking in general, mediolateral (ML) measures have also been investigated, including ML CoM displacement and ML base of support in lower limb amputees who are known to be at greater risk of falling because they are less stable than able-bodied individuals [26–28]. However, while ML CoM displacement has been investigated in relation to stability and falls, this has not been investigated in relation to walking efficiency in lower limb amputees. Given that only vertical CoM displacement is considered unrelated to walking efficiency, which can be explained by energy-conserving theories such as the inverted pendulum model of walking [23], the relationship between ML as well as anterior-posterior (AP) CoM displacement and walking efficiency should be established as this may provide further insight into the biomechanics of efficient walking. In fact, lower limb amputees are known to walk with a wide base of support (BoS) [29,30], which is likely to affect ML CoM displacement and hence may influence walking efficiency and therefore warrants further investigation.

Therefore, the primary aim of this study was to investigate whole-body CoM displacement in all 3 directions in relation to the metabolic cost of walking in amputees with different levels of lower limb amputation as well as in a control group of able-bodied individuals. Also, as there may be a link between metabolic cost and CoM displacement as well as between CoM displacement and the BoS, a secondary aim of this study was to investigate the relationship between BoS and metabolic cost. Finally, as walking speed is also considered an indicator of gait quality, investigating the relationship between walking speed and metabolic cost was a final aim.

2. Methods

2.1. Participant information and study protocol

The required walking data came from another study on walking of amputees and able-bodied individuals, which was recently completed in part by 2 authors of the current study and gives all the details of the data collection protocol [4]. In brief, this involved thirty amputees to form 3 groups of ten unilateral trans-tibial (UTT), ten unilateral trans-femoral (UTF) and ten bilateral trans-femoral (BTF) amputees, as well as ten able-bodied individuals. For the amputees, the study inclusion criteria were: aged eighteen to forty, lower limb amputation as a result of trauma, attending Defence Medical Rehabilitation Centre (DMRC) Headley Court for routine prosthetics treatment, at least 6 months after receiving their definitive prosthesis, no pain consequent to prosthesis usage (minor “discomfort” was acceptable), and capable of walking comfortably for twelve minutes continuously. Study exclusion criteria were: any neuromusculoskeletal pathology (except for the amputation) that may affect the participants’ walking. Each amputee’s definitive prosthesis was chosen and set up on an individual basis, but broadly, amputees were provided with energy storage and return (ESR) feet and micro-processor knees for the trans-femoral amputees. Complete details of the prosthesis prescription for all amputees can be found in the Supplementary materials. Ten able-bodied military individuals needed to be asymptomatic and were also recruited from DMRC Headley Court to provide age- and height-matched control data for comparative purposes (Table 1).

All participants followed the same protocol, which began with 5 min quiet standing while a baseline of oxygen consumption was established using a portable breath analyser (MetaMax 3B, Cortex, Leipzig, Germany). Steady state breathing was verified during data collection by visual inspection of the oxygen consumption data not varying significantly in the final minute compared to the preceding minutes and confirmed retrospectively by comparing the mean and standard deviation of each minute of quiet standing oxygen consumption data to the preceding minute. They then walked for 2 min back and forth along an approximately ten-metre long overground laboratory walking path to establish their self-selected walking speed. Next, they walked for 5 min at their self-selected walking speed to record their oxygen consumption data as well as forceplate data at 1000 Hz (Kistler, Winterthur, Switzerland) and optical motion data at 100 Hz (Vicon, Oxford, U.K.). Due to participant discomfort with the oxygen consumption breathing mask and a failed calibration of the MetaMax system, 3 participants were unable to provide oxygen data and hence their data were excluded from analysis.

A minimum of 5 clean foot contacts were recorded for each limb and analysed separately, with outputs from each gait cycle time-normalised to 100%. A clean foot contact was defined as fully within the boundary of the forceplate. A gait cycle was defined as the time between ipsi-lateral heel contacts, with heel contact being defined by a vertical force greater than 20N applied to the forceplates within the walkway. For all participants, data from the left and right limb were averaged. The mean oxygen consumption from the final minute of both the static trial and walking trial were used to calculate net non-dimensional cost of walking between groups [31,32].

The body was represented as a linked thirteen-segment model consisting of the head, trunk and pelvis, and the left and right upper and lower arm, thigh, shank and foot. Body CoM position was based on the mass-weighted average of body segment parameters scaled according to subject mass and height using

Table 1
Participant demographic information. Values given as mean (S.D.). Note: UTT and UTF groups had fewer than the originally intended 10 participants per group due to problems with the metabolic cost measuring system.

Participants	Mass [kg]	Height [m]	Age [years]
control (n = 10)	78.0 (7.6)	1.82 (0.05)	29 (4)
UTT (n = 8)	88.1 (15.2)	1.83 (0.05)	30 (3)
UTF (n = 9)	88.1 (6.9)	1.80 (0.07)	28 (4)
BTF (n = 10)	86.7 (19.2)	1.81 (0.08)	29 (4)

the anthropometric measures of de Leva [33] and all segments' marker-derived CoM positions.

CoM displacement was defined as the deviation, in all three directions, between the measured body CoM position and the position the CoM would have had if travelling at a constant velocity from the beginning of the gait cycle to the end of the gait cycle (Fig. 1) [34]. BoS was defined as the distance in the ML direction between the left and right fifth metatarsal head marker at heel-strike [28]. It should be clarified that while ML and vertical CoM displacements arise from the body not moving in a straight line, AP displacements arise from the accelerations and decelerations of the CoM in the AP direction over the gait cycle.

2.1.1. Statistical analysis

Linear correlation coefficients were used to determine if there was an association between an individual's metabolic cost and their corresponding CoM displacements. The overall change in CoM displacement was calculated (in each of the vertical, AP and ML directions) for each gait cycle for a subject and was averaged across all gait cycles to give a mean CoM displacement in each of the 3 directions for that subject to be used in the subsequent correlation analysis. p-values are reported if less than 0.05. To determine if differences in CoM displacement were statistically significant between amputee groups, a one-way ANOVA was used with the alpha level set at 0.05, and if the initial test was significant a Bonferroni correction was applied to determine which pair-wise comparisons were different using a paired *t*-test.

3. Results

3.1. Metabolic cost and its correlation with CoM displacement, BoS and walking speed

Metabolic cost was found to increase with amputation level ($P < 0.001$), but was only significantly different when comparing BTF amputees to controls ($P = 0.001$, after Bonferroni correction) and UTT ($P = 0.002$) amputees (Fig. 2).

When comparing metabolic cost to CoM displacement, a significant moderate correlation ($P < 0.001$) was observed with ML CoM displacement (Fig. 3A), but vertical and AP CoM displacement were not correlated to metabolic cost ($P = 0.20$ and $P = 0.77$ respectively) (Fig. 3 B and C). Metabolic cost was found to correlate moderately with BoS ($P < 0.001$) (Fig. 3D) and weakly to walking speed ($P = 0.019$) (Fig. 3E).

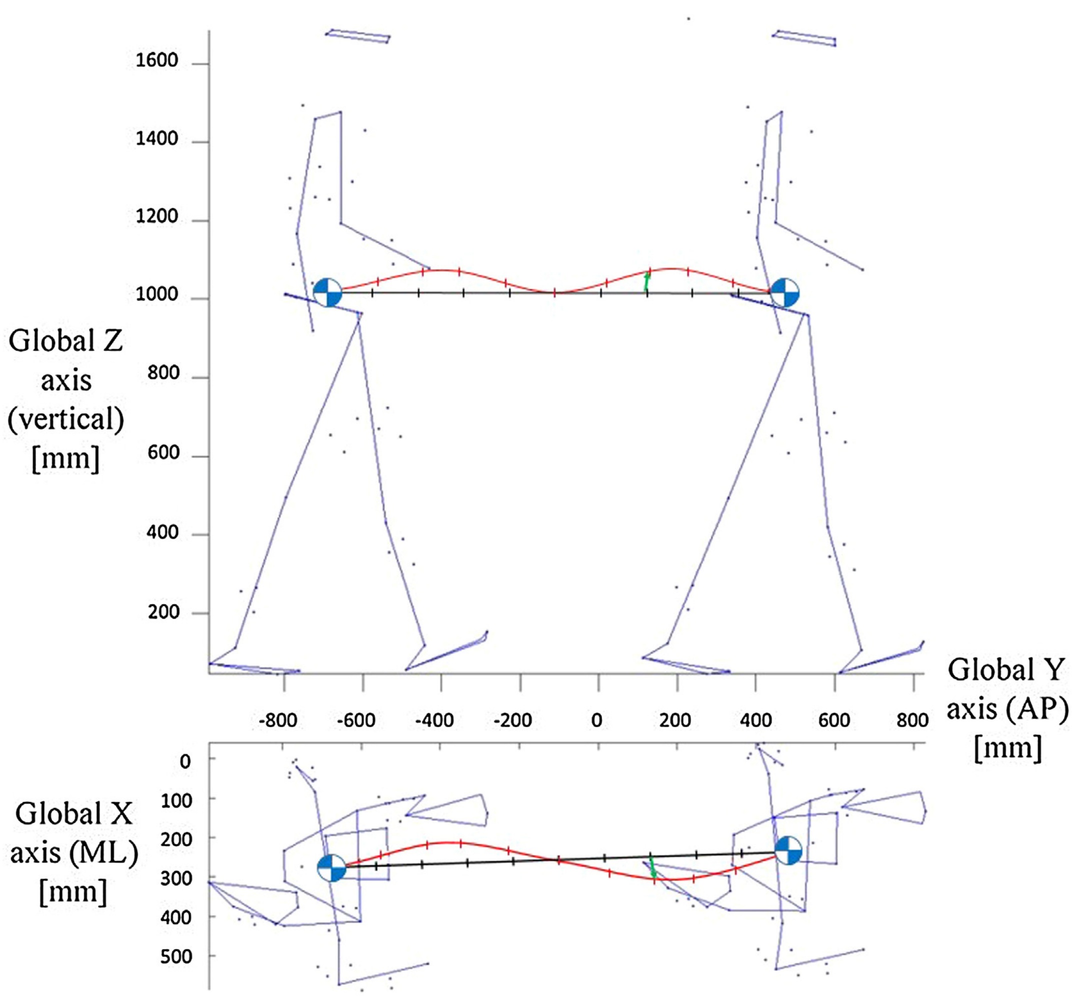

Fig. 1. Definition of CoM displacement. The black lines with equally spaced points represent the trajectory that the CoM would have had if it had travelled from its position at the start of the gait cycle to that at the end at constant velocity. The red lines represent the measured position of the CoM at the same time points. CoM displacement, as defined for this paper, is the distance from a red point to a black point for the corresponding time interval. Note that because all three components of speed vary through the gait cycle that all three components of CoM displacement also vary across the gait cycle. Idealised and exaggerated CoM measured CoM trajectories (red) have been used here to clarify the figure. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

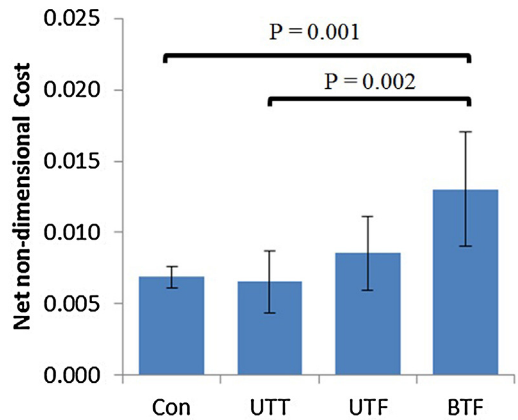

Fig. 2. Comparison of the mean metabolic cost. Note: p-values only given if less than 0.05. Error bars represent standard deviation.

3.2. Group differences in CoM displacement and BoS

ML CoM displacement varied between groups ($P < 0.001$) (Fig. 4), with BTF amputees showing increased ML CoM displacement compared to all other groups. UTF amputees also displayed greater ML CoM displacement compared to controls and UTT amputees. In vertical direction, CoM displacement was also found to be different between groups ($P = 0.001$), with UTT amputees showing increased displacement compared to controls and BTF amputees (Fig. 4). Along the AP axis, no difference in CoM displacement was observed between any of the groups ($P = 0.137$). BoS was also found to increase with amputation level ($P < 0.001$), but was only significantly increased for BTF amputees in comparison to all other groups. The increase in BoS between UTF and controls was approaching statistical significance ($P = 0.019$).

4. Discussion

The primary aim of this study was to investigate whole-body CoM displacement along all 3 axes in relation to the metabolic cost of walking in amputees with different levels of lower limb amputation as well as in a control group of able-bodied individuals. The strongest correlation to metabolic cost was with base of support, with ML CoM displacement showing moderate correlation to cost and walking speed showing a weak correlation to cost.

While ML CoM appeared to be less correlated to metabolic cost than base of support ($R^2 = 0.36$ compared to $R^2 = 0.56$), this can be attributed to two data points in the BTF dataset that lie on the extremes of the dataset in terms of metabolic cost and the normalisation procedure used here. Considering the BTF amputee with the lowest metabolic cost first, while their oxygen consumption at rest and during walking (3.8 and 13.3 ml/min/kg respectively), was in line with the group as a whole (4.1 ± 0.6 and 15.3 ± 2.6 ml/min/kg respectively), this amputee weighed approximately 50 kg more than the group average, which resulted in a substantially lower normalised metabolic cost, despite them walking with a wide base of support and a correspondingly large ML CoM displacement. In contrast to the first individual, the BTF amputee with the highest metabolic cost in the group weighed approximately 15 kg less than the group mean and was approximately 10 cm shorter, but consumed 47% more oxygen at rest and 71% more when walking compared to the group average. If these two individuals are excluded from the group, the correlation between ML CoM displacement and metabolic cost strengthens to 0.55 (from 0.36), which suggests both ML CoM displacement and BoS could be considered equally good indicators of metabolic cost.

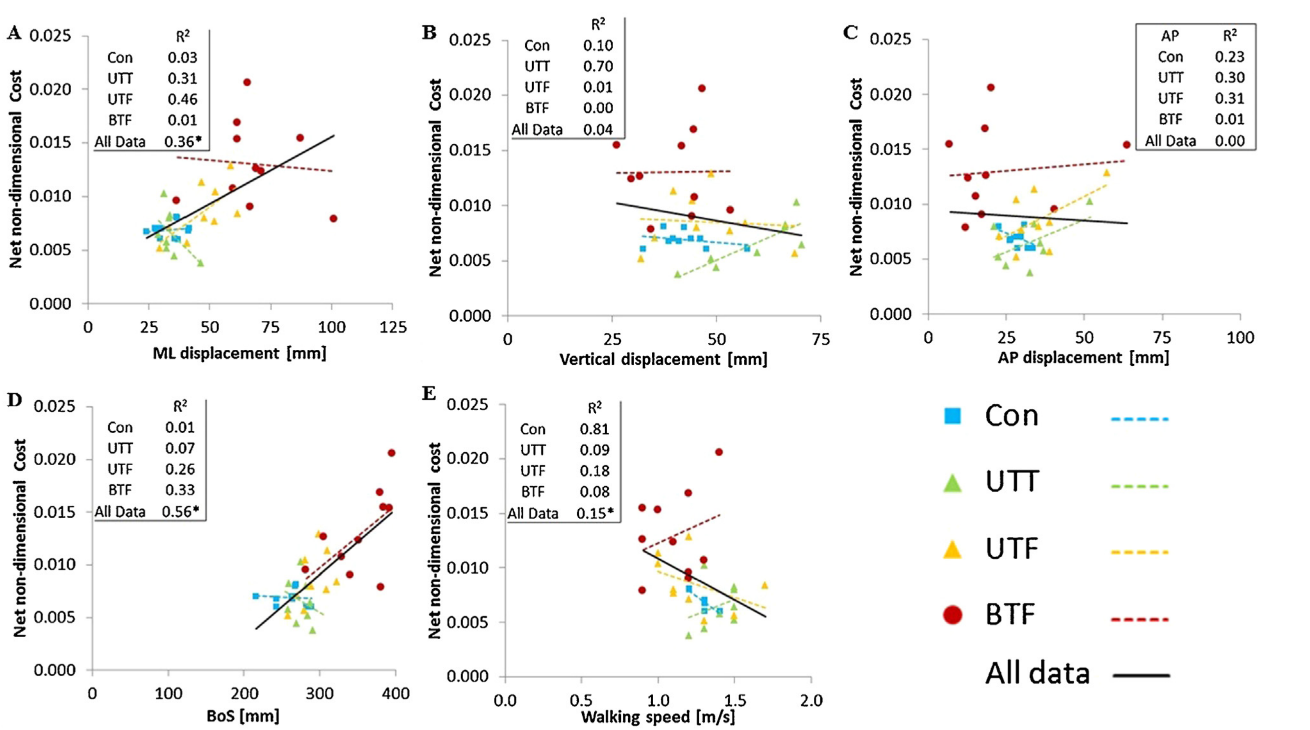

Fig. 3. Correlation of metabolic cost with: ML (A), vertical (B), AP (C) CoM displacement, BoS (D) and Walking speed (E). Note: Dashed lines represent within-group linear regression lines and solid lines represent linear regression lines for all data pooled data. In (D), the trend line for UTF amputees (yellow) lies under the overall trend line for all groups (black). * signifies $P < 0.05$. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

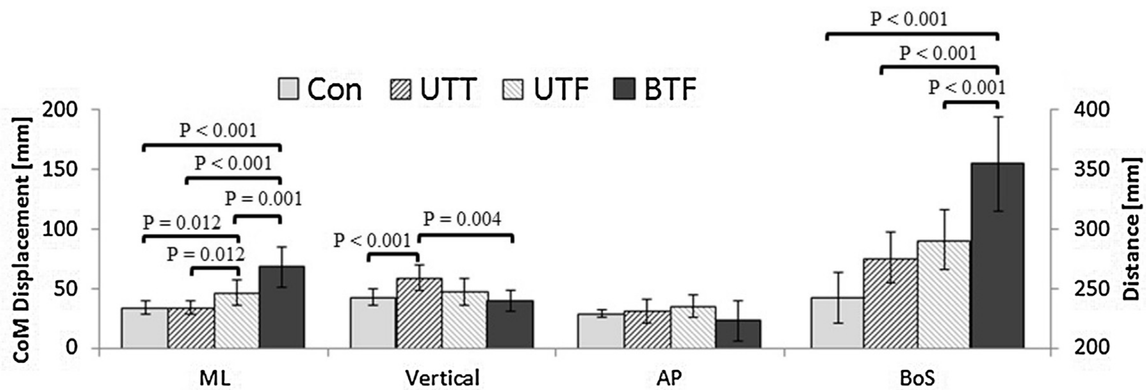

Fig. 4. Across amputee groups, comparison of the mean peak CoM displacements along all 3 axes and of the BoS (right). Note: Error bars represent standard deviation and horizontal bars denote statistically significant differences between specific groups; the right hand y-axis ("Distance") is used for the BoS chart only.

This is supported by the strong correlation between ML CoM displacement and base of support ($R^2 = 0.73$, see Supplementary material), which suggests ML CoM displacement may be a result of the wider BoS, an idea further supported by the consistent moderate to strong within-group correlates shown here.

While lacking statistical power due to the relatively small group sizes and small spread in the measured parameters, it should be noted the within-group correlations for ML CoM displacement and BoS did not always coincide with the overall correlate. However, the UTF and BTF groups demonstrate trends that are representative of the overall dataset and do suggest that in these amputees at least, BoS is a good indicator of metabolic cost. It is worth noting the apparent lack of correlation between ML CoM displacement and metabolic cost in BTF amputees is also attributable to the two data points discussed above, as excluding these points results in a positive correlation coefficient of 0.19 (from 0.01).

Despite base of support correlating most strongly with metabolic cost, this result should be treated as an association between base of support and metabolic cost, rather than one where changing base of support will implicitly change an individual's metabolic cost. Rehabilitation protocols which indirectly reduce base of support will likely show corresponding decreases in metabolic cost, whereas asking amputees to walk with a narrower base of support will likely have the opposite effect, as previous investigations in able-bodied individuals suggests [35]. Despite walking speed being considered an indicator of gait quality and generally being the target of rehabilitation protocols, this was found to have only a weak negative correlation to metabolic cost, a result supported by previous work investigating walking speed as an indicator of metabolic cost ($\rho = -0.38$ in this study, $\rho = -0.36$ in Kark et al.) [16].

Also, given the debate around the relationship between vertical CoM displacement and metabolic cost, two observations from the vertical CoM displacements should be highlighted. The first was the lack of increased vertical CoM displacement with amputation level, and the second was the lack of correlation between vertical CoM displacement and metabolic cost, with the exception of UTT amputees, who demonstrated a greater metabolic cost of walking with increased vertical CoM displacement. However, this observation should be considered specific to this amputee group, as controls and above-knee amputees all displayed no relation between vertical displacement and metabolic cost of walking. These findings complement those from previous studies investigating the relationship between metabolic cost and vertical CoM displacement [17,18], as this study did not aim to change an individual's gait to minimise CoM displacement, but instead correlated an individual's natural displacement with their metabolic cost. The lack of correlation between vertical CoM displacement and metabolic cost supports previous studies that found an increased metabolic cost when deviating from a preferred gait pattern and adds to the evidence suggesting there is no underlying reason why minimising vertical CoM displacement should minimise the metabolic cost. Whilst base of support and M-L CoM displacement both show some correlation with metabolic energy cost it is not possible to say whether the relationship is causal or simply indicative of compensatory adaptations to walking with a prosthesis which affect metabolic energy cost.

The current study has some limitations, which need to be considered as they may impact on the applicability of this study to other work. First, relatively small amputee and able-bodied groups were recruited to form the total participant group, which limits the statistical power of this study and limits the ability to reliably compare trends between groups. However, the aim of this study was to investigate the correlation between metabolic cost and CoM displacement, which is an exercise that allows all 37 participants to be pooled. Naturally, for a correlative study, more subjects in all groups are desirable to provide stronger statistical support for the correlate. Second, all amputations were a result of trauma and may limit the applicability of the results to amputations not caused by trauma, such as peripheral vascular disease. The third limitation of the study is the use of able-bodied anthropometric measures to derive the body CoM position. While the use of device-specific anthropometric measures is ideal, the effect of using able-bodied equations likely resulted in only minor differences in the overall body CoM position, given the relative mass of the prosthesis to the remaining musculature. The final study limitation pertains to the fact only military personnel were tested here. It is likely these individuals are in better physical condition than the average age-matched civilian and as such, this may restrict comparisons to studies assessing civilian cohorts.

5. Conclusions

This study investigated the relationship between CoM displacement, base of support and walking speed to metabolic cost and found the strongest correlation between base of support and metabolic cost, followed by ML CoM displacement. No correlation between vertical CoM displacement and metabolic cost was found. The positive correlations between base of support and ML CoM displacement and metabolic cost suggests increased ML CoM displacement or base of support will reduce walking efficiency. However, this result should be treated as an associative relationship, rather than a causative one. Rehabilitation protocols which indirectly reduce these indicators, rather than vertical CoM displacement will likely show improvements in amputee walking efficiency.

Acknowledgements

The authors would like to acknowledge the Engineering & Physical Sciences Research Council for funding the project (EP/K019759/1).

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2016.09.024.

Conflicts of interest

None

References

[1] J.J. Genin, G.J. Bastien, B. Franck, C. Detrembleur, P.A. Willems, Effect of speed on the energy cost of walking in unilateral traumatic lower limb amputees, Eur. J. Appl. Physiol. 103 (6) (2008) 655–663.
[2] R.S. Gailey, M.A. Wenger, M. Raya, N. Kirk, K. Erbs, P. Spyropoulos, M.S. Nash, Energy expenditure of trans-tibial amputees during ambulation at self-selected pace, Prosthetics Orthot. Int. 18 (2) (1994) 84–91.
[3] C. Detrembleur, J.M. Vanmarsenille, F. De Cuyper, F. Dierick, Relationship between energy cost: gait speed, vertical displacement of centre of body mass and efficiency of pendulum-like mechanism in unilateral amputee gait, Gait Posture 21 (3) (2005) 333–340.
[4] H. Jarvis, A. Bennett, M. Twiste, R. Phillip, J. Etherington, R. Baker, Achieving optimum functional gait in severely injured military amputees: the importance of the rehabilitation programme, Arch. Phys. Med. Rehabil. (Submitted for publication).
[5] Y. Sagawa Jr., K. Turcot, S. Armand, A. Thevenon, N. Vuillerme, E. Watelain, Biomechanics and physiological parameters during gait in lower-limb amputees: a systematic review, Gait Posture 33 (4) (2011) 511–526.
[6] M. Traballesi, P. Porcacchia, T. Averna, S. Brunelli, Energy cost of walking measurements in subjects with lower limb amputations: a comparison study between floor and treadmill test, Gait Posture 27 (1) (2008) 70–75.
[7] K. Parvataneni, L. Ploeg, S.J. Olney, B. Brouwer, Kinematic: kinetic and metabolic parameters of treadmill versus overground walking in healthy older adults, Clin. Biomech. 24 (1) (2009) 95–100.
[8] I.M. Starholm, T. Gjovaag, A.M. Mengshoel, Energy expenditure of transfemoral amputees walking on a horizontal and tilted treadmill simulating different outdoor walking conditions, Prosthetics Orthot. Int. 34 (2) (2010) 184–194.
[9] H. Houdijk, E. Pollmann, M. Groenewold, H. Wiggerts, W. Polomski, The energy cost for the step-to-step transition in amputee walking, Gait Posture 30 (1) (2009) 35–40.
[10] C. Detrembleur, J.-M. Vanmarsenille, F.D. Cuyper, F. Dierick, Relationship between energy cost: gait speed, vertical displacement of centre of body mass and efficiency of pendulum-like mechanism in unilateral amputee gait, Gait Posture 21 (3) (2005) 333–340.
[11] T. Chin, S. Sawamura, R. Shiba, Effect of physical fitness on prosthetic ambulation in elderly amputees, Am. J. Phys. Med. Rehabil. 85 (12) (2006) 992–996.
[12] K.R. Kaufman, J.A. Levine, R.H. Brey, S.K. McCrady, D.J. Padgett, M.J. Joyner, Energy expenditure and activity of transfemoral amputees using mechanical and microprocessor-controlled prosthetic knees, Arch. Phys. Med. Rehabil. 89 (7) (2008) 1380–1385.
[13] L. Torburn, C.M. Powers, R. Guiterrez, J. Perry, Energy expenditure during ambulation in dysvascular and traumatic below-knee amputees: a comparison of five prosthetic feet, J. Rehabil. Res. Dev. 32 (2) (1995) 111–119.
[14] T. Schmalz, S. Blumentritt, R. Jarasch, Energy expenditure and biomechanical characteristics of lower limb amputee gait: the influence of prosthetic alignment and different prosthetic components, Gait Posture 16 (3) (2002) 255–263.
[15] J. Perry, J.M. Burnfield, C.J. Newsam, P. Conley, Energy expenditure and gait characteristics of a bilateral amputee walking with C-leg prostheses compared with stubby and conventional articulating prostheses, Arch. Phys. Med. Rehabil. 85 (10) (2004) 1711–1717.
[16] L. Kark, A.S. McIntosh, A. Simmons, The use of the 6-min walk test as a proxy for the assessment of energy expenditure during gait in individuals with lower-limb amputation, Int. J. Rehabil. Res. 34 (3) (2011) 227–234.
[17] J.D. Ortega, C.T. Farley, Minimizing center of mass vertical movement increases metabolic cost in walking, J. Appl. Physiol. (1985) 99 (6) (2005) 2099–2107.
[18] K.E. Gordon, D.P. Ferris, A.D. Kuo, Metabolic and mechanical energy costs of reducing vertical center of mass movement during gait, Arch. Phys. Med. Rehabil. 90 (1) (2009) 136–144.
[19] J.B. Saunders, V.T. Inman, H.D. Eberhart, The major determinants in normal and pathological gait, J. Bone Joint Surg. Am. 35-A (3) (1953) 543–558.
[20] S.A. Gard, D.S. Childress, The effect of pelvic list on the vertical displacement of the trunk during normal walking, Gait Posture 5 (3) (1997) 233–238.
[21] S.A. Gard, D.S. Childress, The influence of stance-phase knee flexion on the vertical displacement of the trunk during normal walking, Arch. Phys. Med. Rehabil. 80 (1) (1999) 26–32.
[22] D.C. Kerrigan, P.O. Riley, J.L. Lelas, U.D. Croce, Quantification of pelvic rotation as a determinant of gait, Arch. Phys. Med. Rehabil. 82 (2) (2001) 217–220.
[23] A.D. Kuo, The six determinants of gait and the inverted pendulum analogy: a dynamic walking perspective, Hum. Mov. Sci. 26 (4) (2007) 617–656.
[24] J.M. Donelan, R. Kram, A.D. Kuo, Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking, J. Exp. Biol. 205 (Pt. 23) (2002) 3717–3727.
[25] B. McDowell, A. Cosgrove, R. Baker, Estimating mechanical cost in subjects with myelomeningocele, Gait Posture 15 (1) (2002) 25–31.
[26] M.J. Major, R.L. Stine, S.A. Gard, The effects of walking speed and prosthetic ankle adapters on upper extremity dynamics and stability-related parameters in bilateral transtibial amputee gait, Gait Posture 38 (4) (2013) 858–863.
[27] E.J. Beltran, J.B. Dingwell, J.M. Wilken, Margins of stability in young adults with traumatic transtibial amputation walking in destabilizing environments, J. Biomech. 47 (5) (2014) 1138–1143.
[28] D.H. Gates, S.J. Scott, J.M. Wilken, J.B. Dingwell, Frontal plane dynamic margins of stability in individuals with and without transtibial amputation walking on a loose rock surface, Gait Posture 38 (4) (2013) 570–575.
[29] S.M.H.J. Jaegers, J.H. Arendzen, H.J. de Jongh, Prosthetic gait of unilateral transfemoral amputees: a kinematic study, Arch. Phys. Med. Rehabil. 76 (8) (1995) 736–743.
[30] A.L. Hof, R.M. van Bockel, T. Schoppen, K. Postema, Control of lateral balance in walking: experimental findings in normal subjects and above-knee amputees, Gait Posture 25 (2) (2007) 250–258.
[31] M.H. Schwartz, S.E. Koop, J.L. Bourke, R. Baker, A nondimensional normalization scheme for oxygen utilization data, Gait Posture 24 (1) (2006) 14–22.
[32] A.L. Hof, Scaling gait data to body size, Gait Posture 4 (3) (1996) 222–223.
[33] P. de Leva, Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters, J. Biomech. 29 (9) (1996) 1223–1230.
[34] M.H.A. Eames, A. Cosgrove, R. Baker, Comparing methods of estimating the total body centre of mass in three-dimensions in normal and pathological gaits, Hum. Mov. Sci. 18 (5) (1999) 637–646.
[35] J.M. Donelan, R. Kram, A.D. Kuo, Mechanical and metabolic determinants of the preferred step width in human walking, Proc. Biol. Sci. 268 (1480) (2001) 1985–1992.